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THE APPLICATION OF MECHANICAL IMPEDANCE TESTING AND BONE MINERAL MEASUREMENT TO SKELETAL STATUS EVALUATION OF HUMAN CADAVERS

by

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Introduction

Methods and procedures developed by Thompson and Orne (1976), Orne and Mandke (1975), Orne (1974) and Thompson (1973) for the *in vivo* measurement of the bending rigidity (EI) of the human ulna, when coupled with densiometric measurements of bone mineral content (BMC) using the Norland Bone Mineral Analyzer, can be used to evaluate the mechanical properties of the skeletal structure of human cadavers which are being considered for use as surrogates in impact acceleration safety studies. Since the skeleton is the primary supporting structure of the body during impact acceleration, it is essential that candidate cadavers be carefully screened so that test results will be reflective of the desired test population. Especially important is the ability to screen out cadavers with bone structures weakened by extended bed rest, disuse and postmenopausal osteoporosis, etc., unless this reflects the part of the population for which one needs test results. Although it is always possible to excise small lengths of bone (say, the ribs) from the candidate cadavers and subject them to a standard mechanical test procedure, such as three-point bending, nondestructive, noninvasive tests are preferred, especially if the cadaver is ultimately rejected as a test subject. What is really needed is a relatively quick and simple noninvasive, nondestructive way to measure the mechanical strength and rigidity of several long bones in the cadaver to see if the cadaver meets specifications for the impact tests. We shall first briefly describe the techniques which have been developed for making *in vivo* skeletal evaluations, then show how these techniques can be applied to skeletal evaluations of human cadavers.

In Vivo Skeletal Status Evaluation

Numerous pathologies exist in which bones in the body lose calcium (BMC) and become weaker than normal bone. The hip fracture in women with postmenopausal osteoporosis is well known. Bed rest studies have also shown considerable decreases in bone calcium and we can well speculate that paraplegics and quadraplegics who are confined to wheelchairs have weakened bone structures (although we have found nothing in the literature on this subject). Also, partially healed fractures of otherwise normal bone represent structural weaknesses which need to be assessed before the patient can be returned to normal activities.

A natural question arises: How can one determine the *in vivo* breaking strength of skeletal bones without breaking *them*? The answer, of course, is that we can only *infer* the breaking strength from other mechanical or physiological properties which have a high correlation with the breaking strength.

A recent study by Jurist and Foltz (1976) shows a correlation between breaking strength (M) and bending stiffness (EI) of 0.958 and a correlation between bone mineral content (BMC) and breaking strength (M) of 0.947 for 45 embalmed excised human ulnae which were subjected to three-point bending tests. Similar studies recently completed by Borders, Petersen, Orne, Steele and Young (1976) on about 60 dog ulnae, radii and tibiae show a (collective) correlation coefficient between M and EI of 0.80 and M and BMC of 0.84. Grouping the dog bone data anatomically, i.e., by ulna, radius or tibia, appears to give even better correlation coefficients (work in progress) for each bone classification.

The BMC is easily determined by means of the Norland Bone Mineral Analyzer (NBMA) which gives the mineral content in grams per unit of length along the bone with an accuracy within about 2 per cent. The NBMA also gives the width of the bone in the direction of scanning provided it is not "confused" by the presence of other bones very nearby the section being measured. The NBMA is very small and portable and uses a radioactive source of photons whose energy level is fairly low.

The bending rigidity of the human ulna has been measured *in vivo* at the NASA-Ames Research Center by means of the apparatus *shown in* Figure 1. The electromagnetic shaker for generating the harmonic motion and forces transmitted to the subjects' forearm was located on the end of a balance beam. This allowed the magnitude of the preload force acting on the ulna to be adjusted by changing the position of a counterweight. (Preloading could also be accomplished by spring-loading the electromagnetic shaker against some fixed base, if this is found to be desirable in some other laboratory situation.) The purpose of the preload is to stiffen the layer of skin sufficiently so that the driving probe can be as directly coupled to the ulna as possible in order to minimize the filtering through the skin as much as possible. For the probe geometry used, a 600 g preload gives satisfactory results. This is fortunate because this also represents the threshold of pain for most subjects. A description of the electromagnetic shaker and associated electronics is given in Thompson and Orne (1976).

A mathematical model of the mechanical impedance tests is given in Thompson and Orne (1976). An earlier version of the model is given in Orne and Mandke (1975). The model consists of a simply-supported, viscoelastic beam (the ulna, in this case) subjected to a harmonic excitation applied near its mid-span through a layer of skin which is modeled as a viscoelastic, 3-parameter solid (springs and dashpot). The distributed musculature is modeled as a viscoelastic continuum attached to and oscillating with the ulna along its length.

Figures 2, 3 and 4 show typical results of absolute impedance, z, versus forcing frequency and phase angle, ϕ, versus forcing frequency for preloads of 400 g, 500 g and 600 g, respectively. Experimental data points are designated by *'s and the model results are shown by the solid curved line.

The simulations appear quite good up to the first resonant frequency (about 300 Hz). Observe that in the low-frequency range (less than, say, 150 Hz) the impedance curves have a -45° slope (on the log-log plots) indicating an essentially spring-like behavior. If it were not for the fact that the spring constant of the compressed skin layer under the probe is not very large compared to the spring constant of the bone, one could obtain the ulna EI from the simple relation

$$|z| = k/p$$

where p is the forcing frequency (r/s) and k is the static spring constant for a simply-supported beam. For example, if the force is applied at mid-span,

$$k = 48\ EI/\ell^3$$

so

$$EI = k\ell^3/48$$

or

$$EI = |z|p\ell^3/48 \text{ (low frequency)}$$

Unfortunately, the stiffness of the skin must be accounted for and requires the use of the more sophisticated model.

Cadaver Skeletal Status Evaluation

The techniques discussed above for the *in vivo* measurement of bone mineral content and flexural rigidity of the human ulna can be applied with little change to the measurement of the same physiological and mechanical properties of the skeletal structure of human cadavers.

For example, with cadavers there is no limit (within reason) to the amount of preload which can be applied to the various bones. Thus, the coupling of the impedance probe to the bone will be even better in the cadaver than *in vivo*. Since, in many instances, the cadaver will have been embalmed, the skin stiffness at the probe site should be considerably greater than in the *in vivo* case. Again, this improves the coupling of the probe to the bone. Of course, the ultimate coupling is achieved by inserting the probe through a small (1/8" dia) incision in the skin to let the probe bear directly on the bone. This incision does not degrade the structural integrity of the skeleton and, if necessary, can be sewn up after the impedance tests are conducted. This is certainly preferable to excising sections of bones, such as the ribs, for the purposes of conducting standard in vitro mechanical tests. The repair of the skeleton after such bone sections have been removed is possible but experimental results obtained on cadavers modified in this way may be in some doubt if the structural integrity of the skeleton is seriously compromised.

The NBMA is a well-developed piece of portable equipment that can be readily used as is in most laboratory environments. Unfortunately, the apparatus and associated electronics for making the mechanical impedance measurements have been assembled on an ad hoc basis for research purposes and are not miniaturized or packaged for portable use at this time. Preliminary discussions with electronics people in the San Francisco Bay area indicate that the miniaturization and packaging of the equipment and electronics can be accomplished for in the neighborhood of $20,000. Automatic data processing and reduction software can be developed for, say, another $10,000 for a total of $30,000.

In the mean time, it is desirable to perform numerous experiments in which intact cadaver limbs are measured with the present impedance equipment and compared with static tests on the same bones after they have been excised. This data base is necessary before intelligent decisions can be made about the acceptability of individual cadavers for particular impact test purposes.

Also, it would be interesting to correlate the BMC in ribs and other bones which cannot be measured by the NBMA in the intact cadaver to the BMC in long bones such as the ulna, radius, humerous and tibia. Thus, measurement of BMC in, say, the ulna might be indicative of the BMC in other parts of the skeletal structure.

The impedance measuring equipment could possibly also be applied to measuring the stiffness of the rib cage by applying the impedance probe to the sternum. Pilot studies would be needed to determine the best kind of support conditions for the upper torso during these tests and a mathematical model relating individual rib behavior and geometry to the overall fore-aft stiffness of the rib cage would be useful.

Therefore, it appears that there are no significant obstacles to applying existing impedance and bone mineral content measurements to measuring the flexural rigidity and breaking strength of various long bones in the skeleton of cadavers. The suitability of a given cadaver for particular test purposes is determined by the specifications of the test to be run and its measured or inferred mechanical properties. Miniaturization and packaging of the impedance equipment and electronics is needed and an extensive statistical data base for mechanical properties of human cadavers is desirable.

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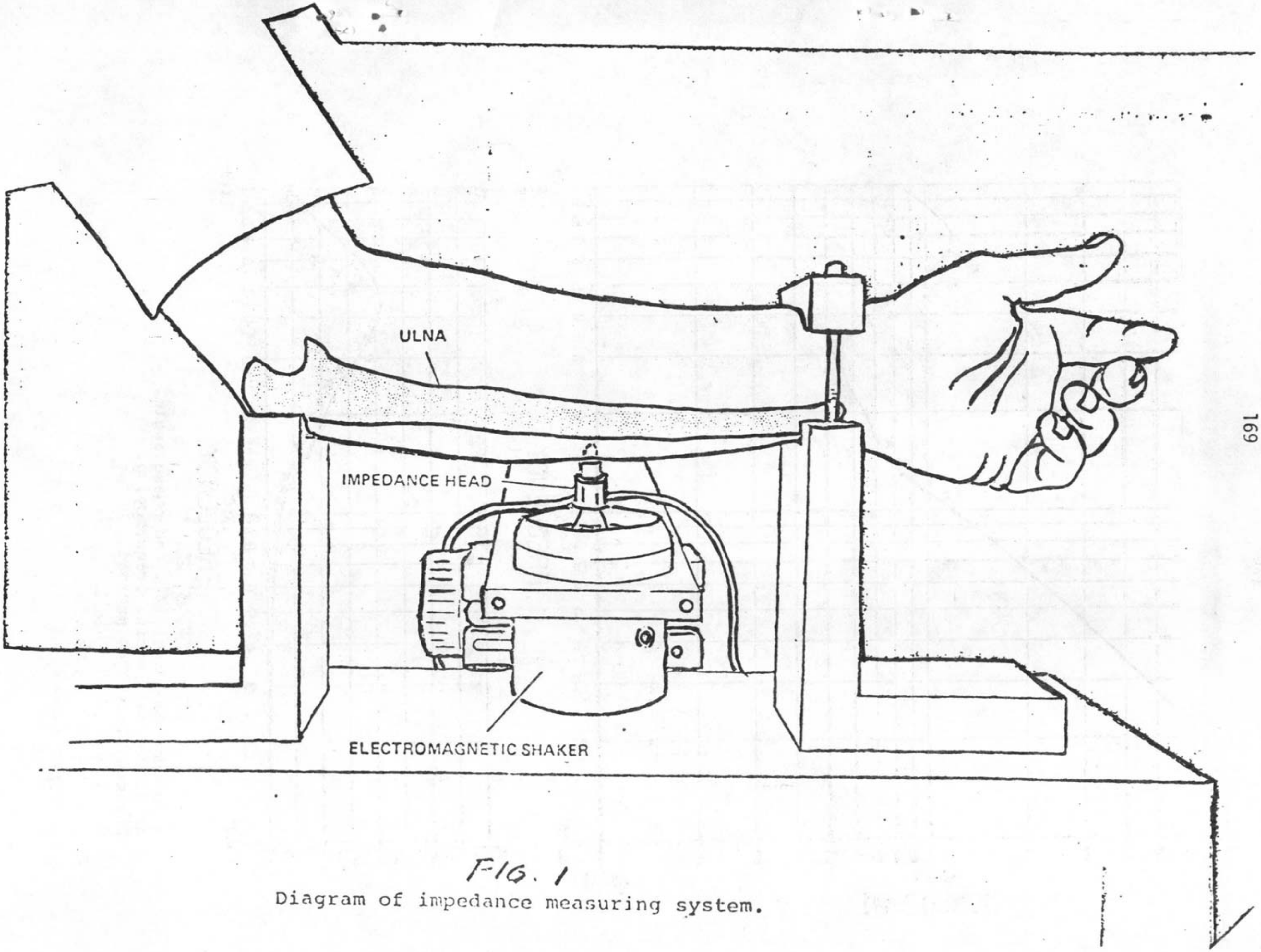

FIG. 1

Diagram of impedance measuring system.

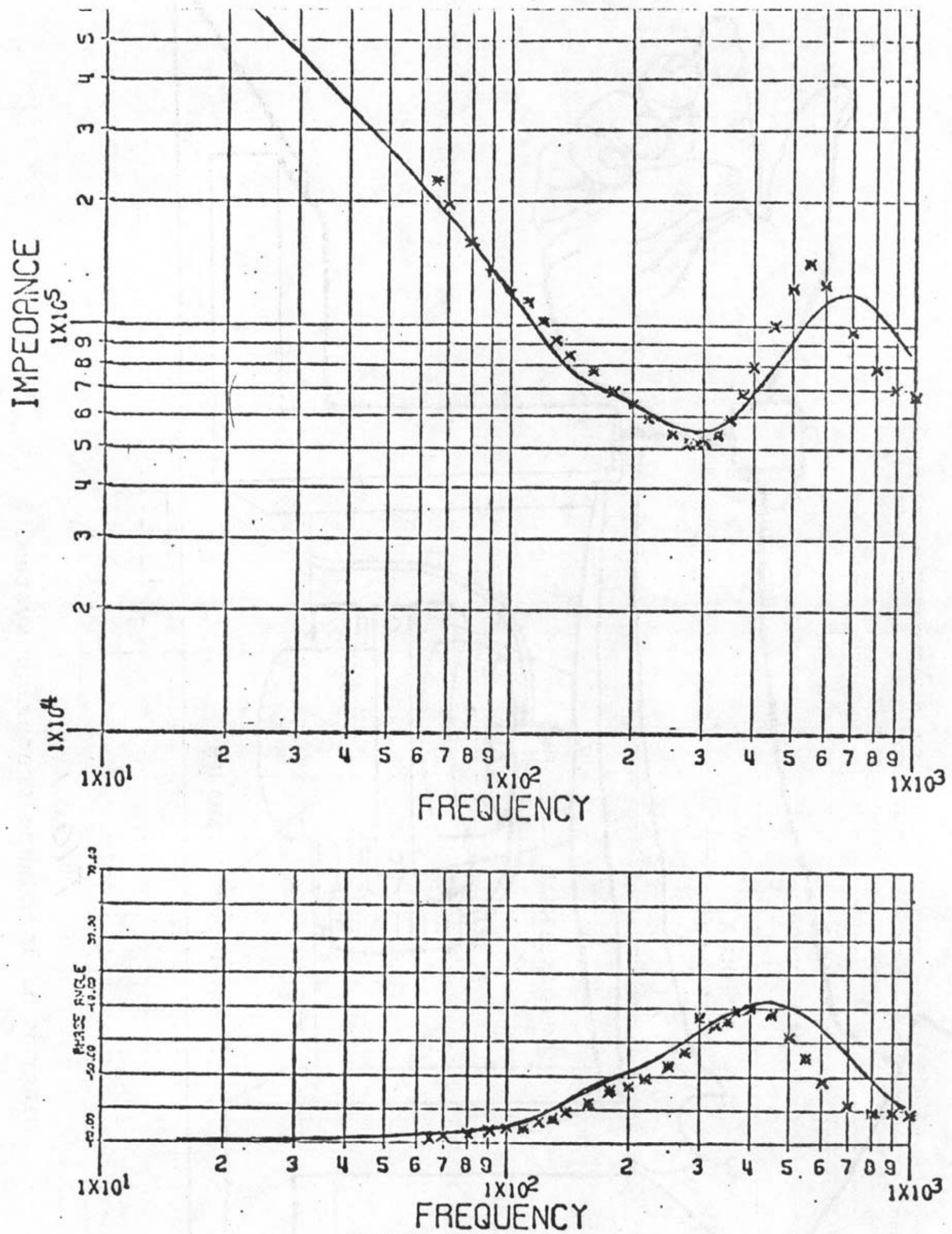

FIG 2.

Absolute impedance, |Z|, and phase angle, arg Z, versus forcing frequency, p. Subject TT: 400 gm preload.

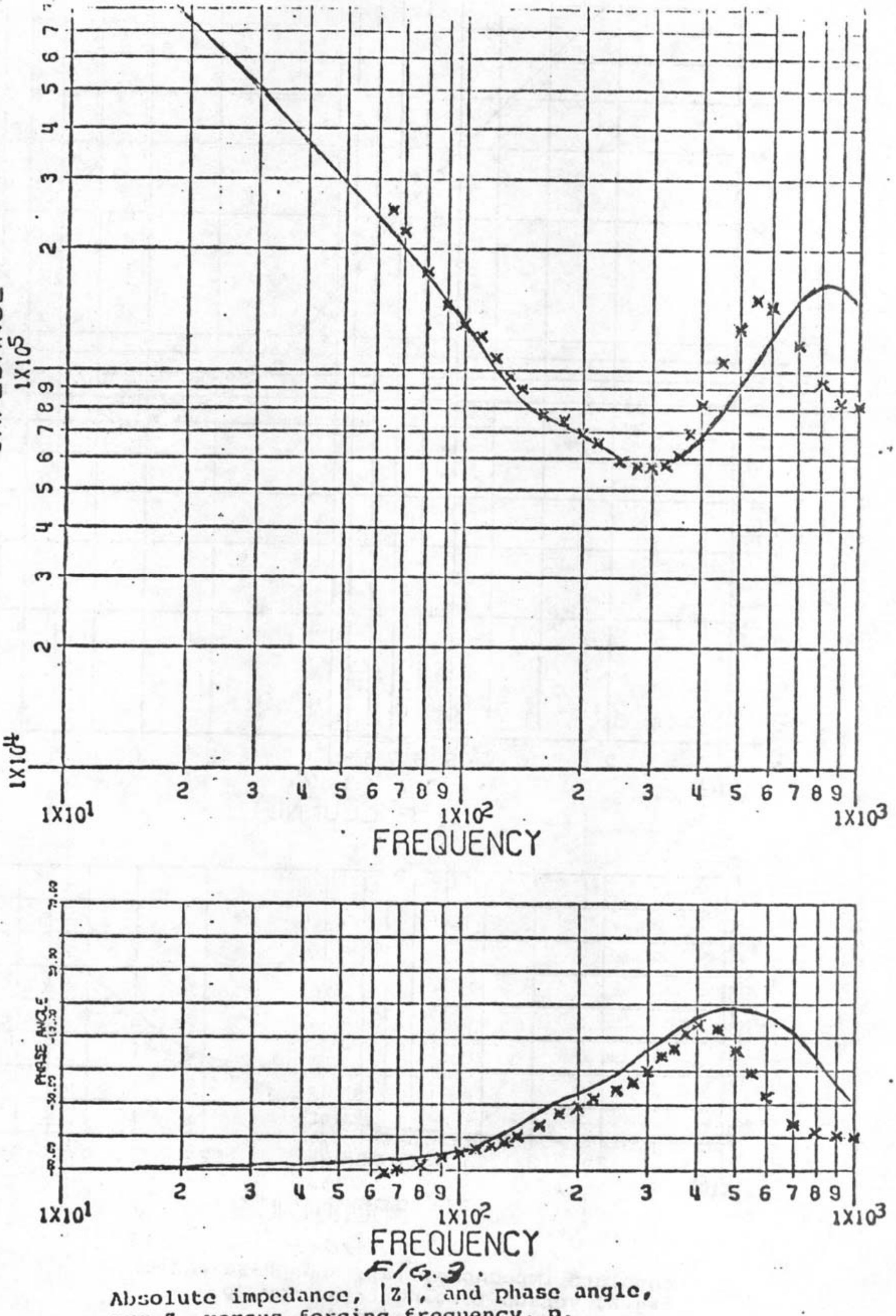

FIG. 3

Absolute impedance, |Z|, and phase angle, arg Z, versus forcing frequency, p. Subject TT: 500 gm preload.

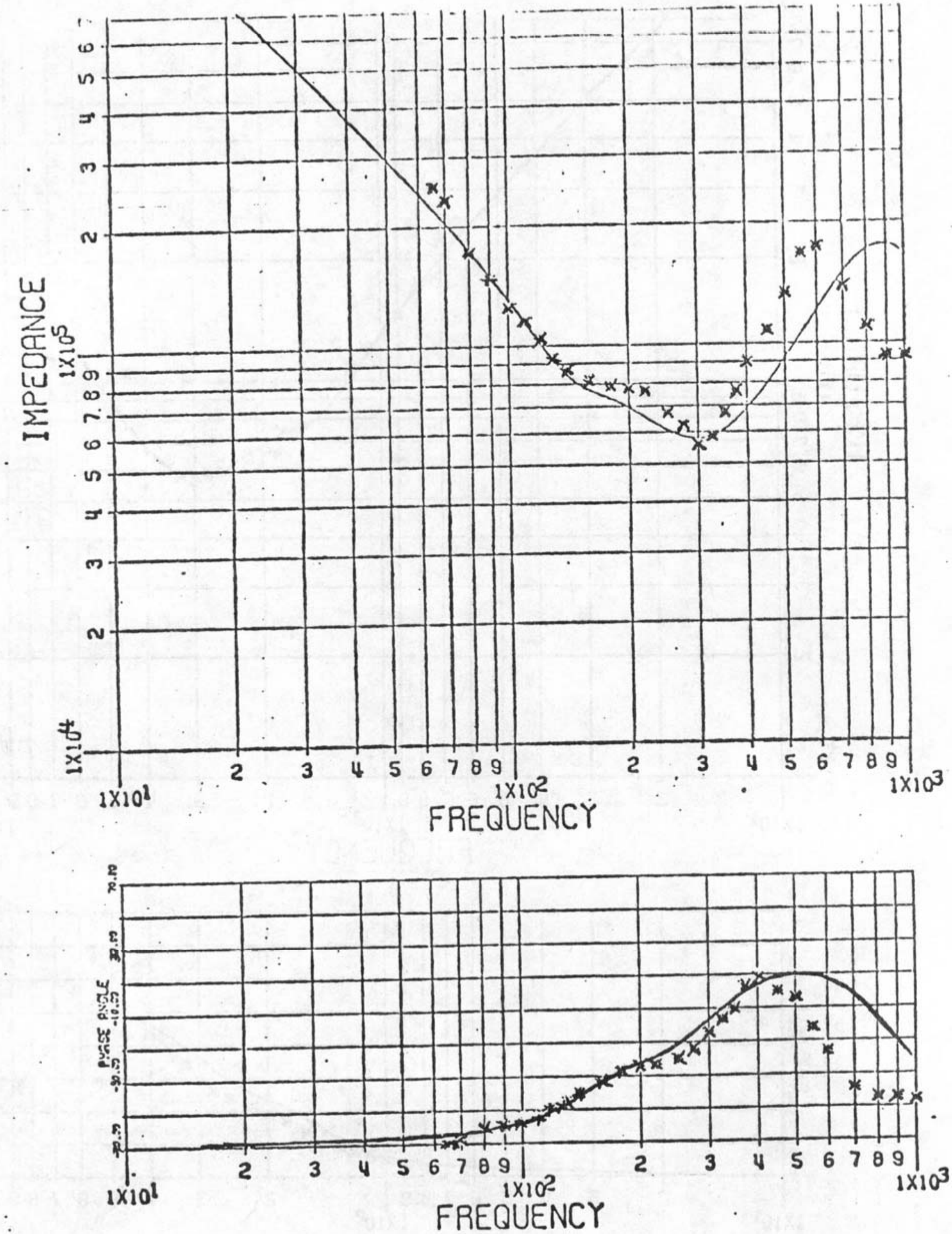

FIG. 4.

Absolute impedance, |Z|, and phase angle, arg Z, versus forcing frequency, p. Subject TT: 600 gm preload.